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(54) **ELECTRODE SPACING IN A
SUBCUTANEOUS IMPLANTABLE CARDIAC
STIMULUS DEVICE**

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CPC **A61N 1/3704** (2013.01); **A61N 1/048**
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CPC A61N 1/08; A61N 1/0476; A61N 1/048
USPC 607/5-7, 115; 600/377, 509
See application file for complete search history.

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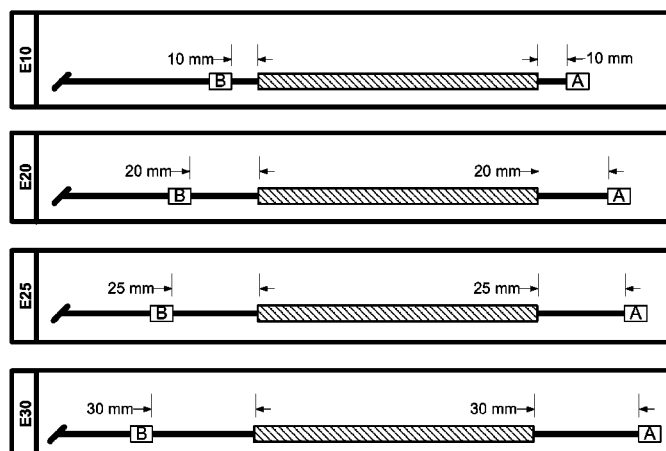
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(57) **ABSTRACT**

Methods and implantable cardiac stimulus devices that
include leads designed to avoid post-shock afterpotentials.
Some examples are directed toward lead-electrode designs
that reduce the impact of an applied stimulus on sensing
attributes. These examples may find particular use in systems
that provide both sensing and therapy delivery from subcuta-
neous location.

20 Claims, 7 Drawing Sheets



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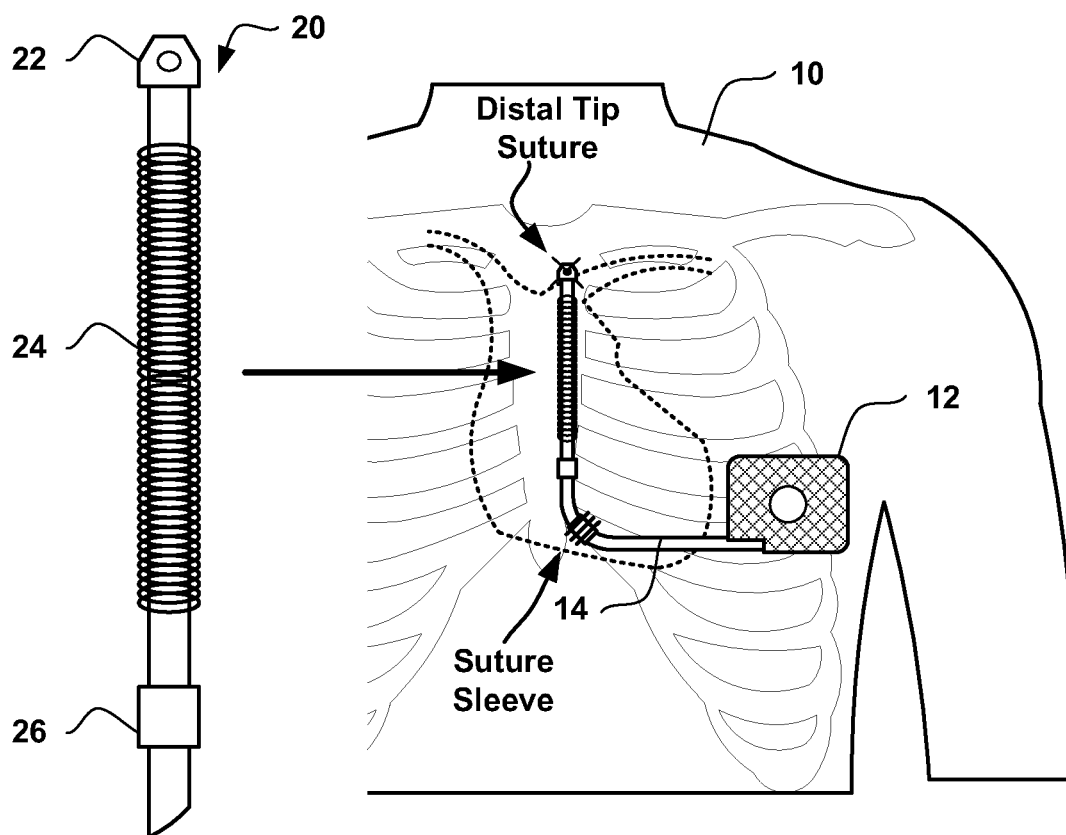
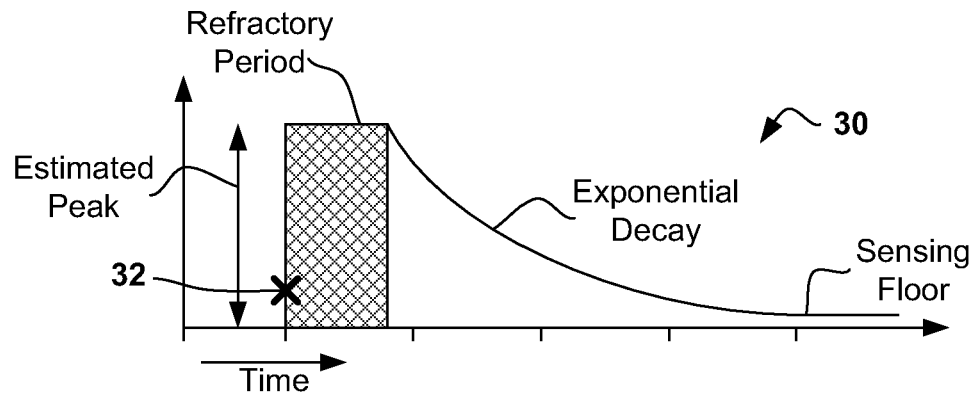
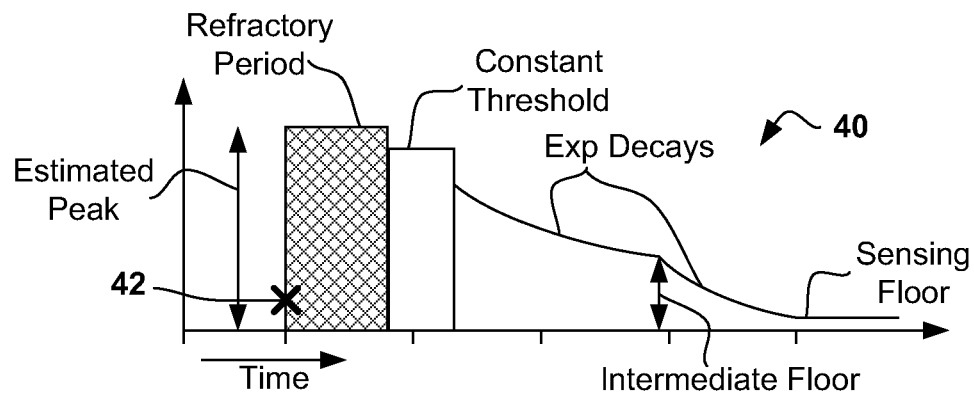


FIG. 1

**FIG. 2****FIG. 3**

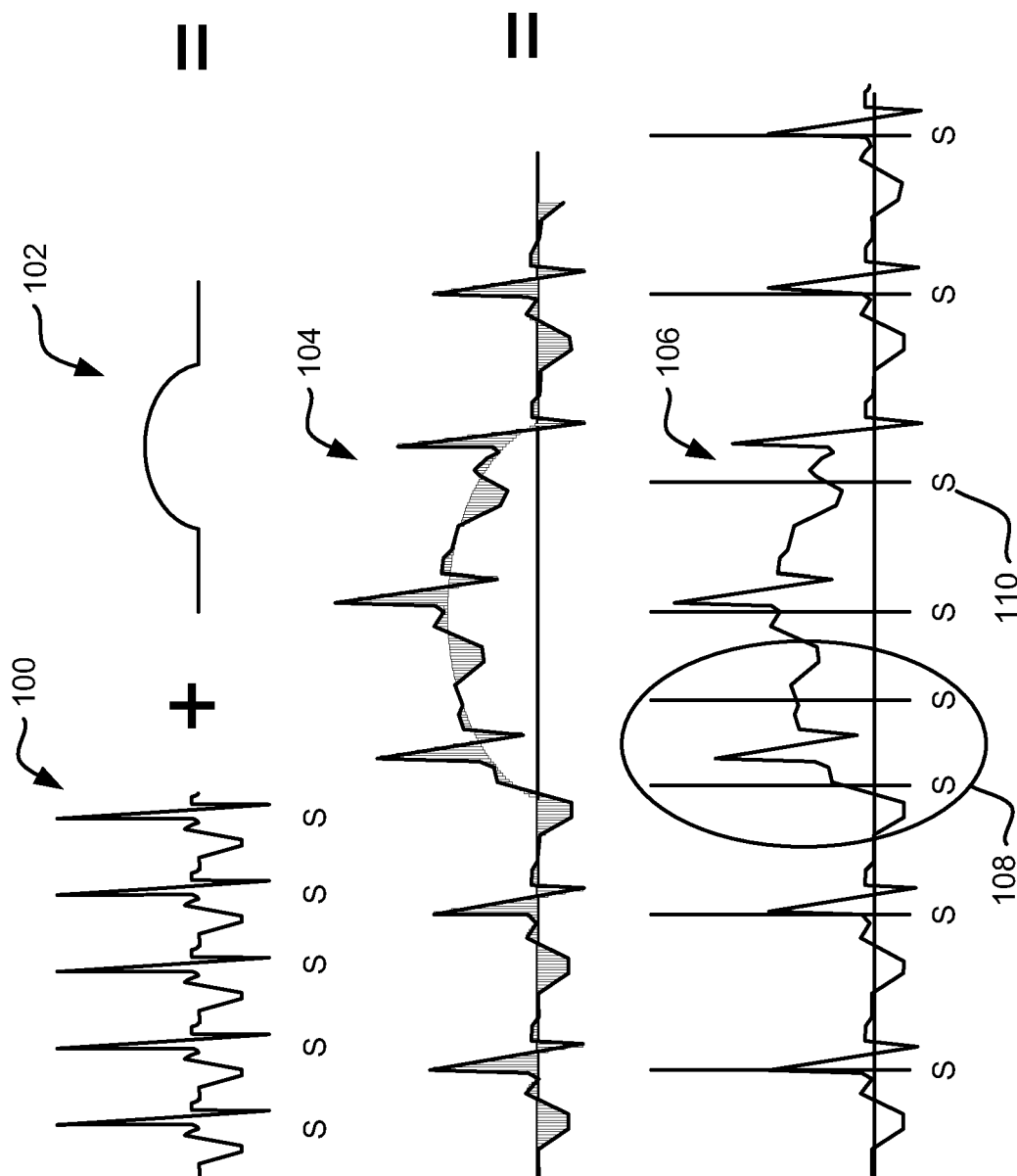


FIG. 4

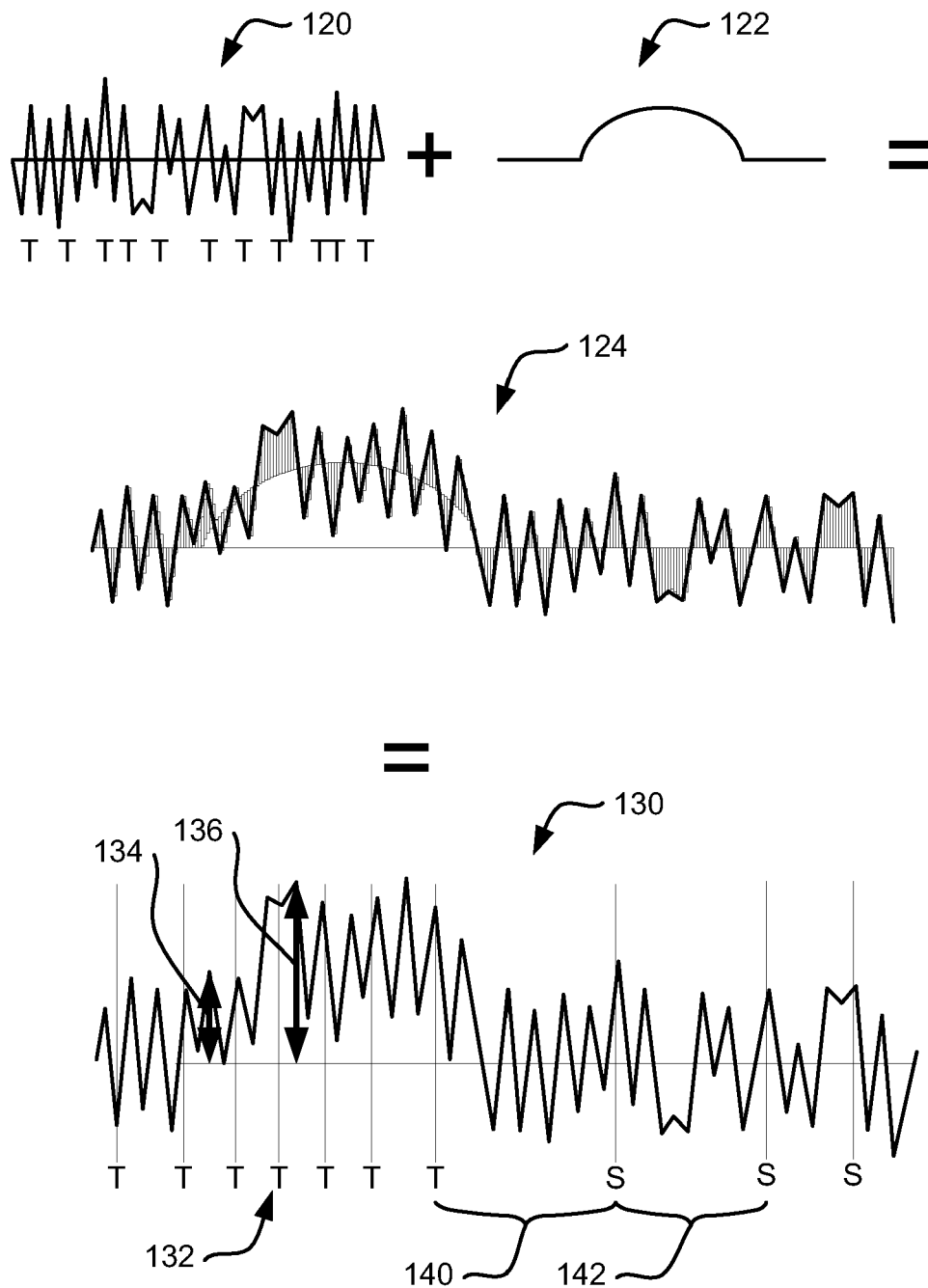


FIG. 5

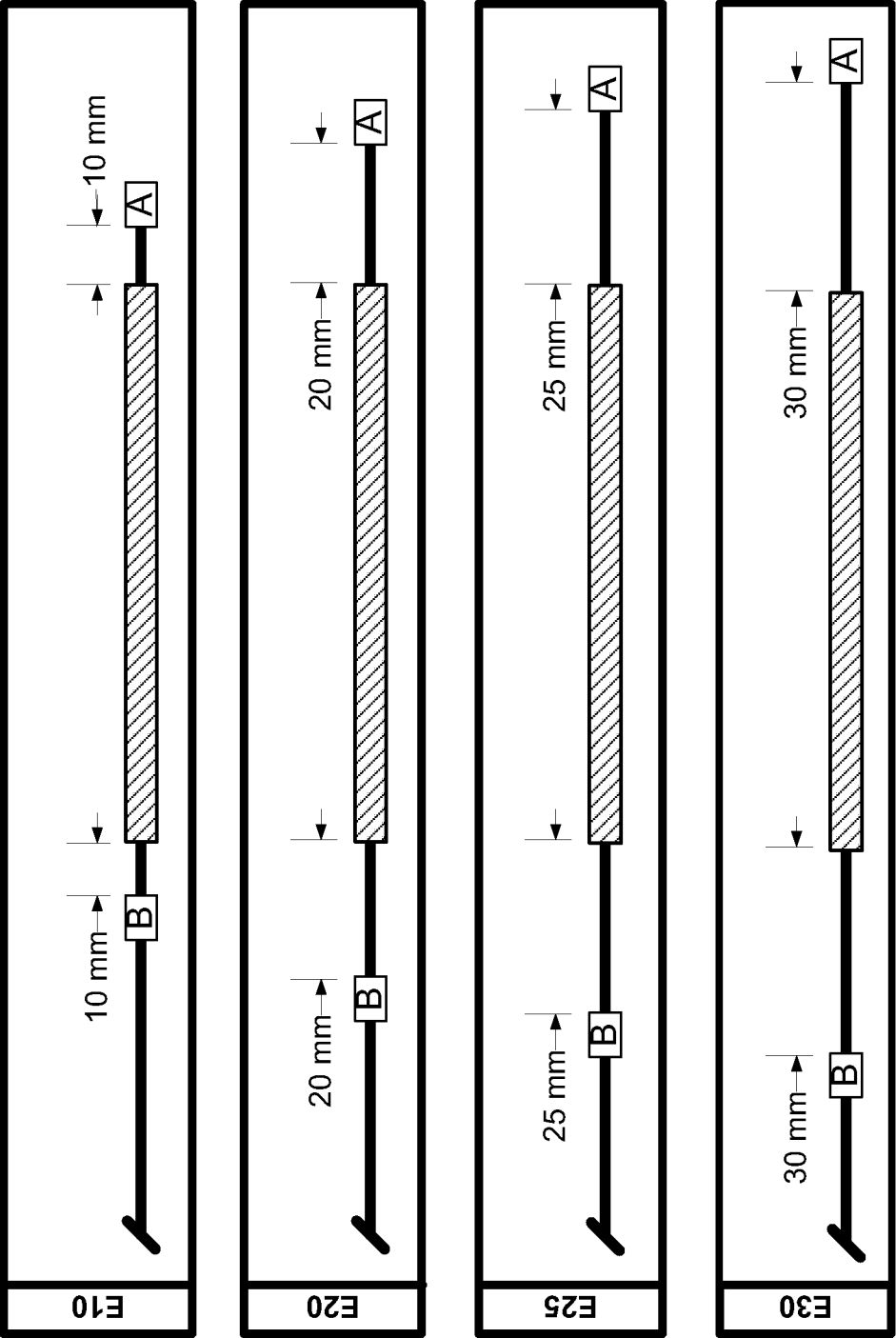


FIG. 6

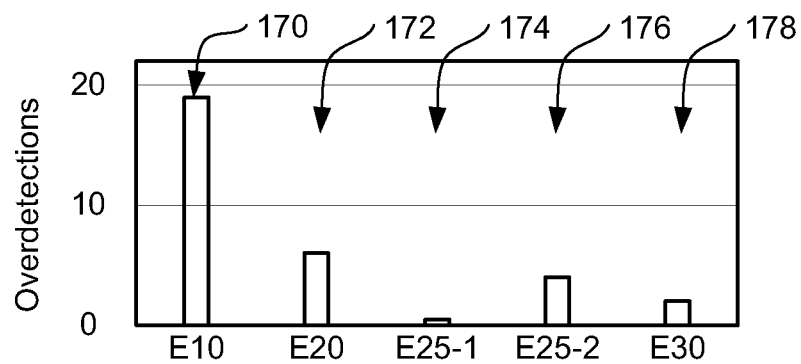


FIG. 7

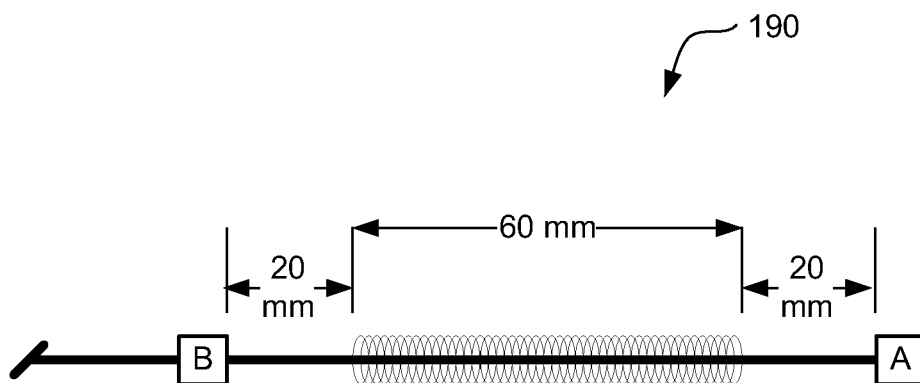


FIG. 8

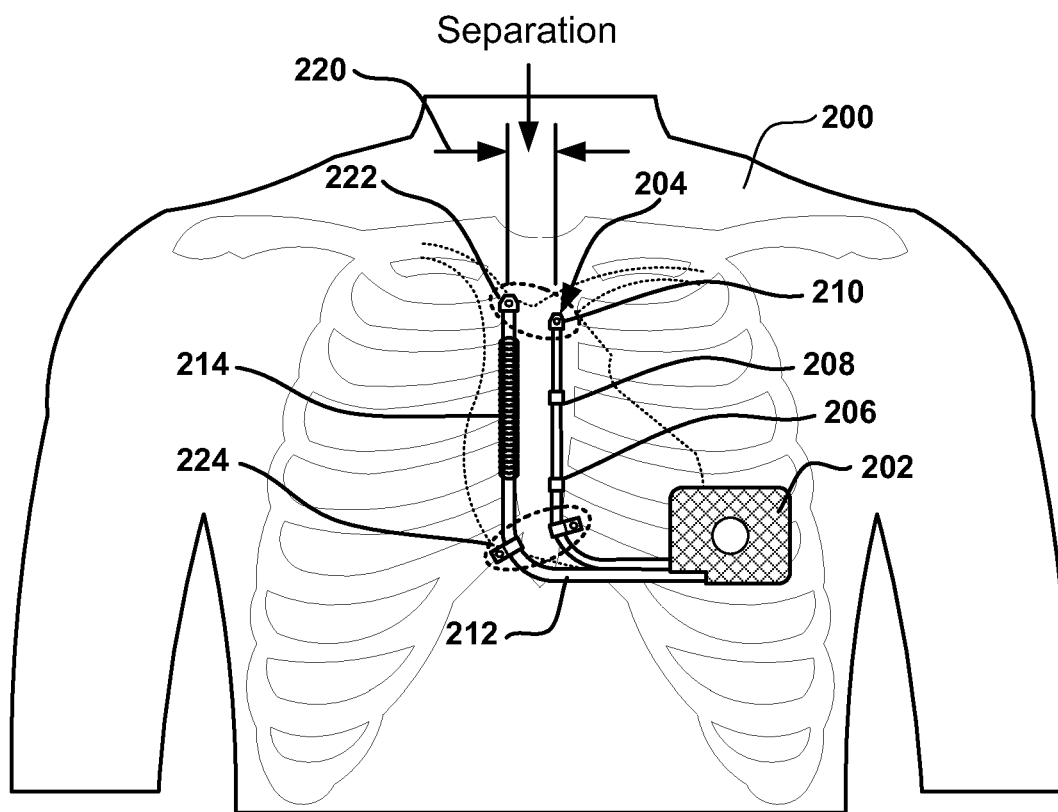


FIG. 9

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ELECTRODE SPACING IN A SUBCUTANEOUS IMPLANTABLE CARDIAC STIMULUS DEVICE

RELATED APPLICATIONS

The present application is a continuation of U.S. application Ser. No. 12/636,569, filed Dec. 11, 2009, now U.S. Pat. No. 8,483,841, which claims the benefit of and priority to U.S. Provisional Patent Application No. 61/122,327, filed Dec. 12, 2008, titled IMPLANTABLE DEFIBRILLATOR SYSTEMS AND METHODS WITH MITIGATIONS FOR SATURATION AVOIDANCE AND ACCOMMODATION, and the disclosure of which is incorporated herein by reference. The present Application is related to U.S. patent application Ser. No. 12/636,575, filed Dec. 11, 2009, published as US Patent Application Publication Number 2010-0152799, now U.S. Pat. No. 8,712,523, titled IMPLANTABLE DEFIBRILLATOR SYSTEMS AND METHODS WITH MITIGATIONS FOR SATURATION AVOIDANCE AND ACCOMMODATION, which also claims the benefit of and priority to U.S. Provisional Patent Application 61/122,327, and is also incorporated herein by reference.

FIELD

The present invention relates to the field of implantable medical devices. More particularly, the present invention relates to subcutaneous implantable cardiac stimulus devices and systems, and leads for use with such devices and systems.

BACKGROUND

Implantable cardioverter-defibrillators (ICDs) are known in the art. Prior devices have included transvenous or epicardial devices. Transvenous devices include leads that reside in blood vessels extending to electrodes placed in the heart. Epicardial devices include electrodes placed on the outside of the heart, usually placed via invasive surgery. New developments in the field include subcutaneous-only systems which lack leads in the vasculature and/or electrodes in or on the heart. Methods and devices configured to respond to new challenges in the subcutaneous ICD space are desired.

SUMMARY

The present invention includes several embodiments that are directed toward lead electrode designs for use in a subcutaneous implantable defibrillator. The following illustrative examples may be embodied in methods of operation, methods of implantation, and/or as implantable components, devices or systems.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 shows an illustrative subcutaneously implanted system relative to the anatomy of a patient;

FIGS. 2-3 are graphic representations of detection profiles for use in detecting cardiac events with an illustrative implantable system;

FIGS. 4-5 graphically simulate post-therapy perturbations of ECG signal and potential effects on event detection;

FIG. 6 shows several illustrative lead electrode assemblies;

FIG. 7 is a histogram of over-detected events during a test of the lead electrode assemblies of FIG. 6;

FIG. 8 shows another illustrative lead electrode assembly; and

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FIG. 9 illustrates an alternative implantation configuration relative to a patient's anatomy.

DETAILED DESCRIPTION

The following detailed description should be read with reference to the drawings. The drawings, which are not necessarily to scale, depict illustrative embodiments and are not intended to limit the scope of the invention. Any references to other patents or patent applications are intended as illustrative of useful methods or devices and are not intended to foreclose suitable alternatives.

FIG. 1 illustrates features of a subcutaneously implanted cardiac stimulus system relative to the anatomy of an implantee 10. The illustrative system includes a canister 12 attached to a lead 14, and both are implanted subcutaneously between the ribcage and the skin of the patient 10. There are no endocardial or epicardial electrodes and no transvenous or intra-thoracic leads in the example shown. The canister 12 is disposed near the left axilla of the patient. The lead 14 extends in a medial direction to approximately the xiphoid and then toward the head of the patient along the left side of the sternum, terminating near the top of the sternum. In an alternative embodiment, one or more leads may be placed in or on the heart in addition to the subcutaneous lead 14, for example, transvenous or epicardial electrodes/leads. The canister 12 may have one or several electrodes, and/or the canister 12 may lack an electrode. Other subcutaneous locations may be used.

As shown in the detail view at 20, the lead 14 includes a distal sensing electrode 22, a therapy delivery coil 24 and a proximal sensing electrode 26. (For the example, "Distal" and "Proximal" refer to position along the lead 14, with the distal electrode 22 being farther from the connection of the lead 14 to the canister 12). The distal sensing electrode 22 may include a suture hole useful for securing the distal sensing electrode 22 to patient tissue during implantation. The canister 12 may also include a suture feature (such as a suture hole in the header) for securing to patient tissue. In some examples, a suture sleeve is used on the lead 14, for example, near the xiphoid, to provide an additional point of fixation or as a substitute for the suture hole on the distal electrode 22. In another example, a suture sleeve may also be placed near the canister 12, either in combination with other fixation points or standing alone. Other fixation structures may be used, without limitation. The lead 14 may include a separate conductor for each electrode 22, 24, 26, and lead 14 may be described as a lead electrode assembly without limiting lead 14 to any particular internal structure or method of manufacture.

Additional illustrative implantation locations and methods of implantation are shown, for example, in commonly assigned U.S. patent application Ser. No. 11/006,291, published as US 2006-0122676 A1 and titled APPARATUS AND METHOD FOR SUBCUTANEOUS ELECTRODE INSERTION, now U.S. Pat. No. 7,655,014, and/or U.S. Pat. Nos. 6,647,292, 6,721,597 and 7,149,575; any of the implantation locations shown in these patents and published applications may be used. Implantation may also be performed by creating incisions at the axilla, xiphoid and termination locations and tunneling therebetween in order to pull the electrode into place using various surgical methods. In one example, an insertion tool is used to tunnel from the xiphoid to the axilla, where an electrode distal end is attached and then pulled back into the xiphoid-axilla tunnel. An insertion tool is then used to tunnel to a termination point superior to the xiphoid along the sternum, with a long suture loop attached at the distal end of the insertion tool and to the electrode distal end. When the

tunnel to the termination point is complete, the suture loop is removed from the distal end of the insertion tool, the insertion tool is removed, and the suture loop is used to pull the electrode into the tunnel to the termination point.

In one alternative example, a system is implanted with the canister **12** disposed approximately anterior of the left axilla with the lead **14** extending posteriorly past the axilla to a location over the posterior region of the ribs of the patient, with the distal end of the lead to the left of the spine of the patient, creating an anterior-posterior stimulus vector between the active canister and the coil electrode **24**. The '292 patent includes certain unitary embodiments that may also be used, as desired. Other locations may be used and, if desired, a transvenous or epicardial sense or therapy lead may be included.

The system may include any suitable components for detection (such as an input ECG amplifier, filtering circuitry, analog-to-digital conversion circuitry), control (such as memory, a microcontroller and various logic circuits, etc.), telemetry (such as an antenna, amplifier, mixer, transceiver, etc.), power supply (any suitable battery technology may be used) and output circuitry (such as switches, high-power capacitors, and charging circuitry). Any suitable materials may be used for the lead **14** and canister **12**. An illustrative example includes a titanium canister **12** having a titanium nitride coating, a polyurethane body for the lead **14** with a silicone sheath thereon, and MP35N conductors in the lead **14** and electrodes **22**, **24**, **26**. Various other materials are known in the art and may be substituted for any of these items. Examples include iridium oxide or porous carbon coatings, platinum or silver electrodes, conductors and/or canister materials, and other materials for the body of lead **14**. Drawn filled tubes and/or other structures or materials can be used as conductors in the lead **14**.

In order to analyze an implantee's cardiac activity, an implanted medical device captures electrical signals using a combination of at least two implanted electrodes. A detection profile is an analytic tool that can be used to identify cardiac events as perceived from the view of the implantable system. The detection profile is compared to the captured electrical signal(s) and, when the captured signal has a greater magnitude than that represented by the detection profile, a detected event is declared.

FIGS. 2-3 show illustrative detection profiles. Additional examples and explanations of detection profiles may be found, for example, in U.S. Pat. No. 5,709,215 to Perttu et al. and/or commonly owned U.S. Provisional Patent Application No. 61/034,938, titled ACCURATE CARDIAC EVENT DETECTION IN AN IMPLANTABLE CARDIAC STIMULUS DEVICE, and U.S. patent application Ser. No. 12/399,901, published as US Patent Application Publication Number 2009-0228057, now U.S. Pat. No. 8,565,878, which claims benefit of and has the same title as the 61/034,938 Application. The illustrative detection profiles shown in FIGS. 2-3 vary in amplitude with time. This is typical but not required. The examples in FIGS. 2-3 are for illustrative purposes and it should be noted that the present invention is not contingent/dependent upon any particular form of detection profile, and other methods of cardiac signal analysis may be used instead.

Referring now to FIG. 2, a most recent detected event is represented by the X at **32**. Following the most recent detected event **32**, the detection profile **30** enters a refractory period during which it does not identify additional cardiac events. The refractory period helps avoid repeatedly sensing the same cardiac event that led to the detected event at **32**. A refractory period may last, for example, from tens to hundreds of milliseconds, depending upon the system design.

Following the refractory period, the illustrative detection profile **30** undergoes a decay to a sensing floor. The decay is shown, in FIG. 2 (and FIG. 3) as an exponential decay, although other decay shapes may be used including stepped, straight line, etc. The exponential decay begins at an amplitude proportional to the "estimated peak." The estimated peak is an estimate made by the implanted system of the electrical strength of detected events. For example, an estimated peak may be set to the peak amplitude sensed during the refractory period, a peak from a previous refractory period, or an average of peaks from a plurality of refractory periods. The estimated peak may be replaced with a fixed value. Some examples start exponential decay at 100% of the estimated peak; others use lesser fractions down to as low as 25%. The "sensing floor" is the minimum sensing threshold or, alternatively, defines the maximum sensitivity.

Another detection profile is shown in FIG. 3 at **40**. Again, starting with a most recent detected event **42**, the detection profile enters a refractory period. The refractory period is followed by a constant threshold period and two exponential decays, first to an intermediate floor and then to the sensing floor. In particular, the detection profile of FIG. 3 provides a number of additional variables that can be manipulated to achieve desired sensing/detection characteristics tailored to individual patients or predetermined conditions. Some examples of these variables include the amplitude of the constant threshold period, the amplitude of the intermediate floor, decay rate(s) associated with the decay periods, and durations for the refractory period, the constant threshold period, and the exponential decays. In one example, durations, thresholds, and decay rates are modified in response to inputs including detected rate, device state, and similarity of estimated peaks for prior detected events. Multiple detection profiles are shown in FIGS. 2-3 in order to indicate that various methods of detection of cardiac events can be suitable.

For each of the examples shown in FIGS. 2-3, it should be appreciated that, for either detection profile **30**, **40** a new detected event can be declared as soon as the refractory period ends. If a sufficiently large signal appears at the end of the refractory period, a new event will be declared. This helps to quickly identify fast cardiac rhythms. It also presents the opportunity for saturation of the input circuitry to cause erroneous high-rate calculations. The detection profiles rely in part on stability of the baseline for accurate event detection. Baseline shifts can interfere with accurate sensing of cardiac events by, for example, creating false detections, introducing error into estimated peaks, and/or causing underdetection.

During animal research on a subcutaneous-only system, it was found that a very slow-decaying post-stimulus voltage was present in the region of the coil **24** following delivery of large amplitude stimuli. In some instances, a biphasic defibrillation stimulus was delivered at 80 joules, with a peak (initial) voltage of about 1350 volts, using a 50% tilt (each phase) and 95 microfarad output capacitance.

In one instance, a chronically implanted canine subject received stimulus in response to a high rate tachycardia. Perturbations were observed in the signal captured in the implanted device following stimulus delivery. Analysis suggested correlation of the perturbations to post-stimulus respiratory distress and associated chest movement. In another instance, the effect was replicated in an acutely implanted porcine subject. Transcutaneous manipulation of the lead following stimulus delivery induced perturbations in the detected waveform. In the porcine subject, the post-stimulus perturbations due to manipulation near the sense and coil electrodes on a lead were observable for over an hour after

stimulus delivery. Prior to the stimulus delivery, however, no perturbation in the signal could be created by similar transcutaneous manipulation. Various types of physical movement may cause perturbations, such as the patient's voluntary or involuntary movement, external force applied by another person or a machine, and distressed or normal respiration. As a result, frequency selective filtering may be only partly effective. As illustration, FIG. 4 simulates a likely superposition of signals that may have led to these observations.

In FIG. 4, the superposition of two signals **100** and **102** is shown. Signal **100** represents a normal cardiac rhythm. The "S" markers indicate beats that would be detected if this signal, by itself, was captured and correctly analyzed by an implantable system. Signal **102** represents a perturbation modeling the kind of perturbation observed due to transcutaneous manipulation of the lead in the porcine model noted above. The observed signal in the canine model was similar. Electrically, the two signals **100**, **102** can be combined by superposition to result in a composite signal as shown at **104**.

If the composite signal **104** is captured by an implantable device, detection accuracy can be reduced, as shown at **106**. In the illustrative analysis at **106**, a detection profile as in FIG. 3 was simulated. An extra detection occurs at **108**, and the detection at **110** does not align with an R-wave. The detection at **110** could be identified as noise by some systems or, in some systems, morphology analysis may mistake the detection at **110** as a malignant cardiac event.

Another simulation is shown in FIG. 5. Here, a lower amplitude tachyarrhythmia is shown and analyzed with more errors than are shown in FIG. 4. A fast rate cardiac signal is shown at **120** (the "T" markers indicate detections identified as indicating tachyarrhythmia). The signal **120** is combined through superposition with a noise perturbation **122**, yielding the composite signal **124**.

Simulated analysis of the composite signal **124** is shown at **130**, using a detection profile as shown in FIG. 3. Large changes in estimated peaks are observed. In particular, the peak amplitude at **134** (due to the cardiac signal alone) is barely half the amplitude at **136** (due to the noise perturbation plus the cardiac signal). In the illustrative example, these peak amplitudes **134**, **136** are used to define the height of the subsequent detection profile. One or more large peak amplitudes, such as peak amplitude **136**, can raise the detection profile to a level that passes over one or more actual cardiac events.

For example, long detection intervals at **140** and **142** result from undersensing caused by the introduction of large peak amplitudes into the data. Due to the long intervals between detected events, the detection method may begin marking one or more detected events as normal/benign rate (indicated by the "S") rather than tachyarrhythmia (indicated by the "T"). Miscalculation of rate can result from such undersensing.

The potential sensing challenges highlighted in FIGS. 4-5 may not prevent accurate/acceptable operation of a system. For example, filtering and further analysis of the captured signals may be performed to overcome any such difficulties. However, mitigation through modification of the lead could provide an additional tool to improve patient outcomes.

Some embodiments of the present invention include specific distal electrode structures for a device as shown in FIG. 1. FIG. 6 shows several configurations for a lead electrode assembly labeled E10, E20, E25 and E30. Each configuration includes three electrodes: sense electrodes A and B, with an 80 mm coil electrode disposed therebetween. The labels indicate the approximate distance between sense and coil electrodes for each configuration. The gaps from B to coil and coil to A were as follows:

{Nominal Values}	B to Coil	Coil to A
E10	10 mm	10 mm
E20	20 mm	20 mm
E25	25 mm	25 mm
E30	30 mm	30 mm

The electrode labeled E10 was the subject of the initial investigation of post-shock manipulation in the porcine model described above. It was shown in that investigation that, with transcutaneous manipulation, false detections could be induced following shock delivery with electrode E10. Subsequent investigation compared performance of the four configurations shown in FIG. 6. Configurations E10, E20, E25 and E30 were subjected to testing in which a defibrillation stimulus was applied through the coil electrode and, next, transcutaneous manipulation was performed on the distal portion of the lead electrode assembly to see if false detections could be induced. Further testing was performed in a porcine subject during an acute implantation study. It was decided that overdetections following stimulus (under transcutaneous manipulation) would be used as a measure of the effectiveness of the variation of the spacing proposed in FIG. 6 in reducing the likelihood of post-stimulus overdetection.

The table above provides nominal values. In the tested models the following distances were used:

For E10, B to Coil, 7.3 mm, Coil to A, 8.2 mm, and Coil length 82 mm;

For E20, B to Coil, 19 mm, Coil to A, 19 mm, and Coil length 78 mm;

For E25, B to Coil, 26 mm, Coil to A, 26 mm, and Coil length 82 mm; and

For E30, B to Coil 30 mm; Coil to A, 29 mm, and A to B, 79 mm.

FIG. 7 shows the results of testing the configurations E10, E20, E25 and E30. As shown at **170**, during a given period of time, nearly 20 overdetections were found during transcutaneous manipulation of the distal lead electrode assembly in the acute porcine study with the E10 electrode. Increasing the interelectrode spacing using the other designs greatly reduced the number of overdetections. As shown at **172**, use of E20 reduced the number of overdetections by more than 50%. A first run with E25 produced no overdetections, as shown at **174**, so a second run was performed, showing some overdetections, as demonstrated at **176**, but less than the number for the E20 electrode. Further reduction was found with the E30 electrode, as shown at **178**.

It is believed that the testing performed and analyzed in FIG. 7 shows improved performance as a result of the selected electrode spacing. Interelectrode spacing of 19 mm and more was tested and found to represent an improvement over spacing of less than 10 mm. The improved performance made the system less susceptible to incorrect sensing in the post-shock context. Confirmatory testing was performed and showed that additional electrodes having sense electrode-defibrillation electrode gaps of 19 and 20 mm continued to avoid the perturbations and oversensing following shock delivery with physical manipulation, when compared to that which was observed with the 10 mm nominal gap.

It should be noted that the phenomenon identified in post-stimulus sensing with the subcutaneous-only systems shown above is unlikely to be the same as "stunning" sometimes associated with the effects of high voltage stimulus on myocardial tissue, for example, in transvenous systems. Stunned myocardial tissue typically demonstrates contractile dysfunction and reduced detected amplitude for a temporary

period following certain high amplitude stimuli. Because the tissue adjacent the subcutaneous electrodes used in therapy delivery for the systems tested is not myocardial tissue, the observed phenomena is likely of different origin. While systems could operate in the face of this perturbation by using filtering and/or noise detection algorithms, the designs identified herein may enhance post-therapy sensing.

In another illustrative example, one or the other of the sensing electrodes can be omitted. For example, some embodiments make use of a distal tip sensing electrode, but omit the proximal sensing electrode. In other embodiments, the proximal sensing electrode is included, but the distal tip sense electrode is omitted. In yet another example, cardiac signal analysis is performed in the following manner: a default sensing vector is chosen making use of one of three available vectors (referencing FIG. 1): distal sensing electrode **22** to canister **12**; proximal sensing electrode **26** to canister **12**, or distal sensing electrode **22** to proximal sensing electrode **26**, with the default vector selected using, for example, a combination of amplitude and signal-to-noise ratio. Continuing the example, if a vector using the electrode on canister **12** is the default vector, for a period of time after defibrillation is delivered using the canister **12** and coil **24**, an alternate vector may be used. In this example, the post-stimulus vector is selected to omit the shock delivery electrodes **12**, **24**.

Another design is shown in FIG. 8. A distal portion of a lead electrode assembly is shown at **190**. Here, a shorter coil electrode, having a length of about 60 mm, is provided with gaps of nominally 20 mm (actually 19 mm or more) on either side to sense electrodes A and B. The shorter coil electrode may make the implantation location shown in FIG. 1 easier, particularly for patients having a shorter sternum, for example, small adults and/or children. The coil electrode may be shorter still, for example, down to 40 mm or less, if desired.

Another illustrative embodiment is shown in FIG. 9. In this example, an implantee **200** has received an implantable cardioverter defibrillator system that is implanted entirely subcutaneously, generally over the anterior portion of the ribcage. A canister is provided as shown at **202** along the inframammary crease at approximately the left axilla. A sensing lead electrode assembly is shown at **204** as extending vertically along the left side of the patient's sternum and includes several sensing electrodes shown at **206**, **208**, **210**. A stimulus lead electrode assembly is shown at **212** and includes a stimulus coil electrode as shown at **214**. The stimulus lead electrode assembly **212** is shown extending to the right of the patient's sternum. In this example, the sensing lead electrode assembly **204** is implanted such that it is spaced by at least a predetermined separation **220** from the stimulus lead electrode assembly **214**. The separation **220** may be in the range of about 19 mm or more.

In the example of FIG. 9, each lead electrode assembly may be implanted by dissecting a tunnel. By adequately spacing the tunnels, the lead assemblies **204**, **212** may be spaced from one another sufficient to assist sensing operation. Suture sleeve(s) and/or other anchoring features may be used to secure either or both lead assembly **204**, **212** with the desired separation **220**, as shown at **222**, **224**. Special implant tools may be used to create this separation **220**. For example, an insertion/dissection tool having parallel prongs may be used to establish predicable separation **220**.

In another example similar to that of FIG. 9, the separate tunnels on either side of the sternum may be used in the opposite fashion shown, with the shock coil lead **212** being to the left of the sternum and the sensing electrode lead **204** placed on the right of the sternum. This would keep the

sensing electrodes outside of the electrical field formed between the shock coil **214** and the canister **202** during defibrillation therapy delivery.

Those skilled in the art will recognize that the present invention may be manifested in a variety of forms other than the specific embodiments described and contemplated herein. Accordingly, departures in form and detail may be made without departing from the scope and spirit of the present invention.

What is claimed is:

1. An implantable defibrillator system for subcutaneous placement comprising:

an implantable canister housing operational circuitry for sensing electrical cardiac activity and delivering electrical stimulus to correct arrhythmia; and

a lead electrode assembly comprising:

a proximal connector end for coupling to the implantable canister;

a distal end having a distal tip with a distal tip electrode;

an elongated body extending between the proximal connector end and the distal end; and

a defibrillation electrode;

wherein the distal tip electrode is spaced from the defibrillation electrode by a portion of the elongated body having a length of at least nineteen (19) millimeters, with no other electrode therebetween;

wherein the implantable canister is configured to enable insertion of the lead electrode assembly for operation therewith.

2. The implantable defibrillator system of claim 1 wherein: the lead electrode assembly further comprises a proximal sensing electrode proximal of the defibrillation electrode such that the defibrillation electrode is disposed between the proximal sensing electrode and the distal tip electrode; and

the proximal sensing electrode is spaced from the defibrillation electrode by a portion of the elongated body having a length of at least nineteen (19) millimeters, with no other electrode therebetween.

3. The implantable defibrillator system of claim 2 wherein the spacing between the defibrillation electrode and the distal tip and proximal sensing electrodes is sized such that, following delivery of a defibrillation therapy to a patient, afterpotential is minimized.

4. The implantable defibrillator system of claim 2 wherein the spacing between the defibrillation electrode and the distal tip and proximal sensing electrodes is sized to prevent motion artifact following delivery of defibrillation therapy.

5. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized such that, following delivery of a defibrillation therapy to a patient, afterpotential is minimized.

6. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized to prevent motion artifact following delivery of defibrillation therapy.

7. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized such that, following delivery of an 80 Joule defibrillation therapy to a patient, afterpotential is minimized.

8. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized to prevent motion artifact following delivery of an 80 Joule defibrillation therapy.

9. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal

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tip electrode is sized such that, following delivery of a defibrillation therapy having a peak voltage of about 1350 volts to a patient, afterpotential is minimized.

10. The implantable defibrillator system of claim 1 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized to prevent motion artifact following delivery of a defibrillation therapy having a peak voltage of about 1350 volts to a patient.

11. An implantable defibrillator system for subcutaneous placement comprising:

an implantable canister housing operational circuitry for sensing electrical cardiac activity and delivering electrical stimulus to correct arrhythmia; and

a lead electrode assembly comprising:

a proximal connector end for coupling to the implantable canister;

a distal end having a distal tip;

an elongated body extending between the proximal connector end and the distal end;

a defibrillation electrode; and

a proximal sensing electrode proximal of the defibrillation electrode;

wherein the proximal sensing electrode is spaced from the defibrillation electrode by a portion of the elongated body having a length of nineteen (19) millimeters up to twenty-nine (29) millimeters, with no other electrode therebetween;

wherein the implantable canister is configured to enable insertion of the lead electrode assembly for operation therewith.

12. The implantable defibrillator system of claim 11 wherein:

the lead electrode assembly further comprises a distal tip electrode at the distal end such that the defibrillation electrode is disposed between the proximal sensing electrode and the distal tip electrode; and

the distal tip electrode is spaced from the defibrillation electrode by a portion of the elongated body having a length of nineteen (19) millimeters up to twenty-nine (29) millimeters, with no other electrode therebetween.

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13. The implantable defibrillator system of claim 12 wherein the spacing between the defibrillation electrode and the distal tip and proximal sensing electrodes is sized such that, following delivery of a defibrillation therapy to a patient, afterpotential is minimized.

14. The implantable defibrillator system of claim 12 wherein the spacing between the defibrillation electrode and the distal tip and proximal sensing electrodes is sized to prevent motion artifact following delivery of defibrillation therapy.

15. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the proximal sensing electrode is sized such that, following delivery of a defibrillation therapy to a patient, afterpotential is minimized.

16. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the proximal sensing electrode is sized to prevent motion artifact following delivery of defibrillation therapy.

17. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the proximal sensing electrode is sized such that, following delivery of an 80 Joule defibrillation therapy to a patient, afterpotential is minimized.

18. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the proximal sensing electrode is sized to prevent motion artifact following delivery of an 80 Joule defibrillation therapy.

19. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the distal tip electrode is sized such that, following delivery of a defibrillation therapy having a peak voltage of about 1350 volts to a patient, afterpotential is minimized.

20. The implantable defibrillator system of claim 11 wherein the spacing between the defibrillation electrode and the proximal sensing electrode is sized to prevent motion artifact following delivery of a defibrillation therapy having a peak voltage of about 1350 volts to a patient.

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